Image Quality of CR Mammography

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Summary

This paper discusses the technical concepts of Computed Radiography (CR) mammography, and the imaging performance that users can expect using this modality.

Introduction

Computed radiography is an accepted method of performing general digital radiography for applications requiring moderate productivity, resolution, and dose performance. Because of certain technical limitations, however, CR systems do not offer the resolution performance of the state-of-the-art flat panel image receptors used in full field digital mammography (FFDM).

CR Mammography works similarly to CR as used in the radiology department. A special cassette, containing a charge storage phosphor, is used in place of the conventional x-ray screen-film cassette. It is exposed to radiation using standard techniques, and then the technologist feeds the cassette through a scanner. The scanner is illustrated in Figure 1.

The scanner contains a laser beam that is directed to a small location on the phosphor. When the laser beam strikes the phosphor, it stimulates the emission of light, which is detected using two photosensitive elements, one on each side of the plate. The amount of light that is emitted at a given location on the cassette is proportional to the amount of x-ray radiation that that location was exposed to during the acquisition. By scanning the laser beam in a raster motion across the phosphor, and by recording the quantity of light emitted at each location, the scanner assembles a digital image.

This entire scanning process takes about one minute per cassette. The digital image is available after the scanning and is presented to the technologist on a preview computer monitor for image quality review and acceptance.

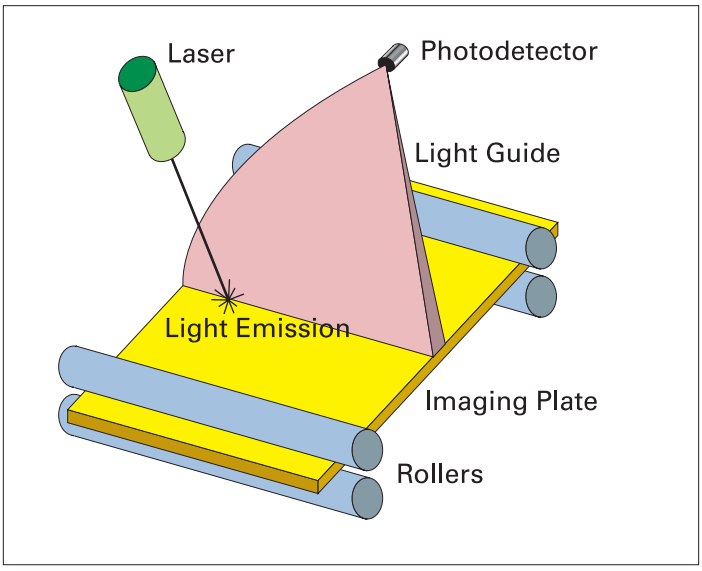

Figure 1. Schematic of CR reader

Technical limitations of CR mammography

High resolution and good dose performance are two prerequisites for a mammography screening system, and CR has difficulty achieving both of these requirements. Understanding why this is requires delving into the physics of computed radiography.

CR is one of a class of image receptors known as indirect conversion detectors. Other detectors that use this technology are amorphous silicon cesium iodide flat panel detectors such as employed in the GE Senographe series FFDM systems. Indirect conversion detectors involve the diffusion of light as part of the x-ray detection process, and this degrades the spatial resolution and makes it difficult to achieve good dose efficiency.

Figure 2 shows this process for an indirect conversion flat panel detector. When an incident x-ray is absorbed in the scintillation layer, typically cesium iodide, the energy of the x-ray is converted into a cloud of low energy visible light photons. These photons diffuse through the scintillator, until they are collected by the photodiodes, which form the top of each pixel. The diffusion is the reason for the resolution loss- many pixels receive a signal from even a single incoming x-ray. Physicists would describe this by saying that the point spread function is

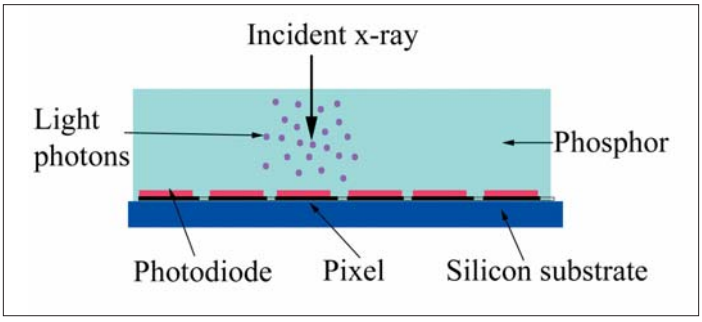

Figure 2. Indirect conversion detectors work by absorbing x-rays, which give off light scintillations that are detected by photodiodes

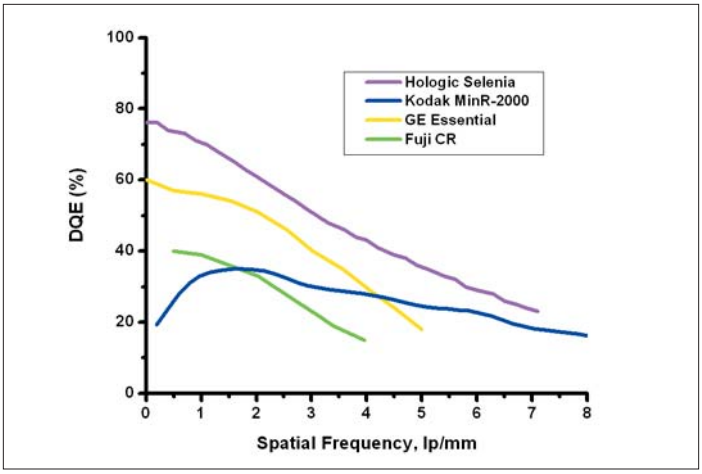

Figure 3. Dose efficiency as shown by the Detective Quantum Efficiency for Fuji CR, compared to screen-film (Kodak), cesium iodide (GE), and selenium detectors (Hologic)

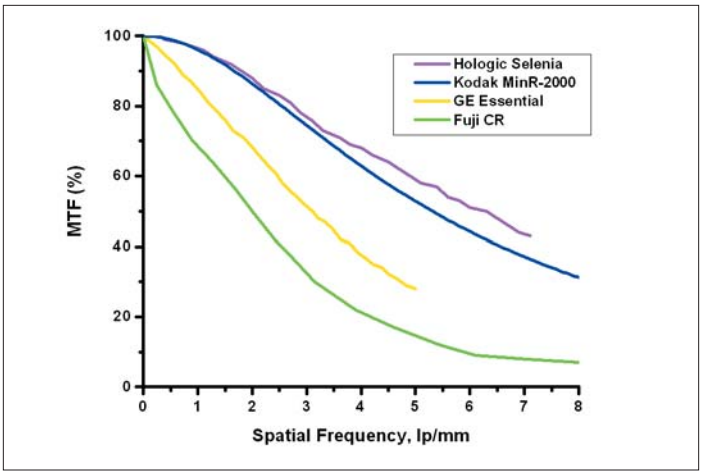

Figure 4. Resolution performance as shown by the Modulation Transfer Function for Fuji CR, compared to screen-film (Kodak), cesium iodide (GE), and selenium detectors (Hologic)

broad. It is important to understand that the inherent resolution of the system is determined by this light spread, and not by the pixel pitch. In particular, making the pixels smaller will not create an image with superior resolution, and we shall see the implications of this for CR Mammography.

Another characteristic of indirect conversion systems relates to dose efficiency. Dose efficiency in its simplest form relates to how efficiently the detector uses the radiation that impinges upon it. Clearly, if the radiation passes through the detector without being absorbed, it does not contribute to the image while it did contribute to increasing the patient dose. The solution to this is to make the detection layer thicker, to increase the probability of absorption of the x-ray. This increases the dose efficiency. Unfortunately, when this is done, the light can diffuse further from the point of absorption of the x-ray and the resolution degrades.

So far we have talked generically about indirect conversion detectors. Now let us look at the specific technical challenges faced by CR. CR uses indirect conversion. Just like indirect conversion flat panel detectors, the x-ray is absorbed in the phosphor, however with CR, there is no immediate scintillation of light. Rather, the light energy is stored in the screen by exciting electrons into a metastable energy state. When the CR plate is put into the CR reader and scanned with the laser beam, the laser beam provides enough energy to release the metastable electrons and we get stimulated emission of light at the locations where we had the metastable state.

CR screens cannot be made too thick, because then we would get poor spatial resolution. They suffer from light diffusion similarly to indirect conversion flat panel detector. In addition, scatter of laser light inside the CR phosphor provides another source of image blurring that further degrades image sharpness.

So we cannot make the screen too thick because our resolution suffers, however if we make the screen thinner then the x-rays will have a significant probability of not being absorbed and the dose efficiency will suffer.

In general for CR detectors, one cannot have both good dose efficiency and high resolution. We shall see that designers of CR systems had to balance these tradeoffs, and have developed systems with resolution significantly inferior to screen-film and competing digital systems. While CR dose efficiency can be made similar to screen-film, it is significantly inferior in comparison to flat panel FFDM detectors.

Dose efficiency

The dose efficiency of a detector can be characterized using the Detective Quantum Efficiency, or DQE, curve. The DQE of various mammography detectors are shown in Figure 3. The horizontal x axis represents line pairs per mm (lp/mm)- which is the inverse of spatial size. Higher lp/mm represent smaller size objects in the image. The DQE is plotted on the vertical y axis. A DQE of 100% would represent perfect dose efficiency. It can be seen that all detectors have the characteristic that the performance of dose efficiency gets poorer and poorer as the objects get smaller, i.e. larger lp/mm.

The dark blue line represents a common screen-film system, Kodak MinR-2000, and can be considered as a reference point. Fuji CR Mammo is shown in green. For the largest objects it has a DQE exceeding screen-film, however above about 2 lp/mm, corresponding to objects of size approximately 500 microns, its DQE, or dose efficiency, is inferior to screen-film. The yellow curve is the DQE for the GE detector that is based on cesium iodide, which also becomes inferior to screen-film for the highest spatial frequencies, or identically, for the smaller image objects.

The curve in purple is the selenium image receptor used in

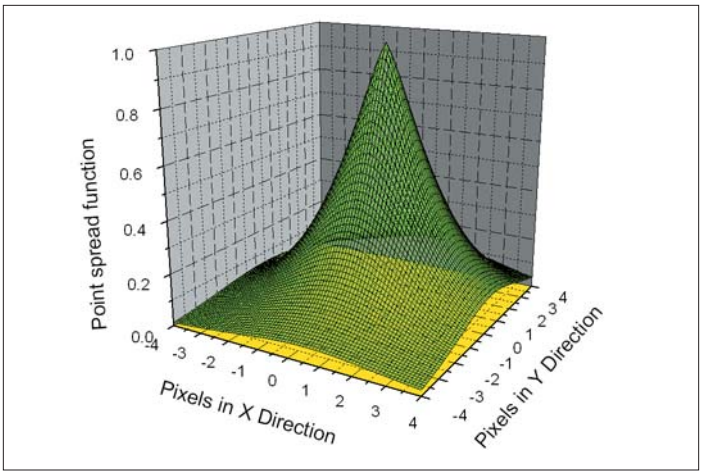

Figure 5. Point spread function for CR mammography system

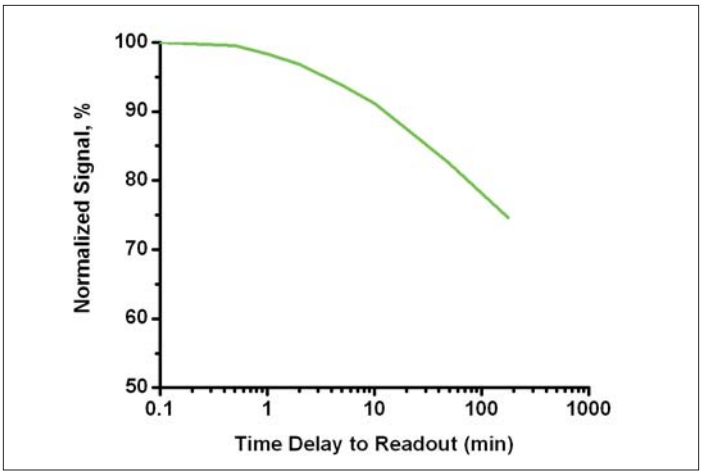

Figure 6. Signal decay from delay between exposure and readout for CR detectors

the Hologic Selenia system. It can be seen that its DQE exceeds that of CR, cesium iodide, and of screen-film by a significant margin. This advantage corresponds to superior image quality at a matched dose or the ability to image at lower doses and still maintain acceptable images.

Resolution performance

Resolution performance can be illustrated using a curve known as the Modulation Transfer Function, or MTF. The MTF for the Fuji CR detector is shown in Figure 4. MTF is plotted similarly to DQE. The x-axis represents lp/mm, and the y-axis shows the contrast when imaging line pairs at a given lp/mm. Like DQE, MTF performance degrades as the objects get smaller and smaller.

The MTF curves show even poorer CR performance compared to screen-film. Its resolution is inferior to analog mammography and to cesium iodide systems.

Selenium image receptors have considerably superior resolution performance compared to both cesium iodide and CR. The direct conversion technology employed in selenium detectors provides both high dose efficiency and superior resolution characteristics. Selenium detectors are the only ones that offer superior resolution performance than screen-film. It is one of the reasons why systems based on these detectors have microcalcification visibility superior to film.

Resolution and image size considerations

It is informative to look at the resolution characteristics of CR using another metric, known as the point spread function. The point spread function shows the detector response from a single incident x-ray and characterizes the resolution in the x,y detector coordinate system. The point spread function can be calculated from the MTF curves. Figure 5 shows this for the 50 micron pixels used in the Fuji CR readout.

What can be seen from this figure is that there is significant signal in many pixels for a single incident x-ray. This represents the inherent resolution limit of CR mammography. Relative to the pixel receiving the largest signal (just under the incoming x-ray), pixels as far away as 3 pixels to the left and right are receiving significant signals. Another way of stating this is there is significant signal in perhaps 49 (7 in the X times 7 in the Y direction) pixels for every x-ray.

One could make the argument that using 50 micron pixels is not optimal. The system could use much coarser pixels without affecting observable image quality because resolution is determined by the broad point spread of the laser beam during readout.

As the pixels are made smaller, the image file sizes increase, and this introduces many difficulties such as longer transfer time, larger PACS storage requirements, and inability to display all the pixels on monitors. If the detector has high intrinsic resolution, such as direct conversion selenium detectors do, then there is value in small pixels. The CR system has all the disadvantages associated with small pixels, with none of the gain in image resolution. This sentiment was echoed by some researchers[7] who noted that "Combined, these findings suggest that a … larger pixel size may be a better match for the optical and electrical properties of this system."

Signal loss, decay over time

CR has another limitation that is unique to its technology. We have seen that the CR reader will stimulate the emission of the stored energy as light. However, the light is actually being emitted all the time, even before the screen is put into the reader. This can be thought of as phosphorescence. Immediately following the x-ray exposure, there is a slow, but steady, emission

of the stored light signal. The longer the interval between the exposure and the reading, the smaller will be the resultant image signal that the CR reader can detect, and so the poorer the image. Figure 6 shows some experimental results on the Fuji CR Mammography screens.

If only ten minutes transpires from exposure to readout, approximately 10% of the signal will have been lost. Thus, in a mammography screening environment with a patient throughput of one every fifteen minutes, the first image taken in the four-view mammogram series will have suffered the equivalent of having received 10% less radiation and will exhibit increased noise. It is important to remember that the patient received the full radiation; it is just that the image quality does not reflect this.

Clinical performance

There have been a number of scientific presentations that attempt to estimate the performance of CR mammography in a variety of imaging tasks.

The paper *Microcalcification detectability for four mammographic detectors: flat-panel, CCD, CR, and screen/film* by Fetterly and Schueler found that CR performed poorer than screen-film, whereas a flat panel detector was superior to both screen-film and CR. They also found the minimum size of detectable microcalcifications to be poorest for CR, and best for flat panel.

The paper *Digital luminescence mammography (CR) versus full-field digital mammography (DR): A phantom study* by R. Schulz-Wendtland et. al. found significantly higher detectability rate of lesions for flat panel full-field digital mammography compared the CR high resolution digital phosphor storage plate in an experimental setting.

The paper *Comparison of full-field digital mammography (FFDM) and CR Mammography: Physical imaging properties and contrast-detail characteristics* by Ideguchi et. al. found that the full field digital mammography system had superior physical imaging properties and contrast-detail characteristics compared to a CR digital mammography system with pixel size of 50 microns.

These results are not surprising given the relatively poor resolution and imaging performance of a CR detector, compared to direct-to-digital mammography detectors.

Conclusion

The resolution and dose performance of CR mammography is inferior to screen-film and to flat panel detectors. The performance of direct conversion selenium-based systems is significantly superior to CR.

In addition, CR offers no productivity advantage compared to screen-film, which is enjoyed by flat panel detectors. And finally, CR is unable to do dynamic imaging, such as tomosynthesis, and this will limit its usefulness and applicability for future applications.

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